Two-dimensional ultrasonic computed tomography of growth bone

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Abstract—**This study deals with the 2-D ultrasonic qualitative and quantitative imaging of child bone. The inverse problem is linearly and non-linearly solved via a Born–based procedure involving minimization of the discrepancies between measurements and modeling data. Inversions of experimental data are presented.**

Keywords: Ultrasonic Computed Tomography, Bone imaging, Born approximation, iterative distorted method

I. INTRODUCTION

Ultrasonography is the main first-line imaging technique used to diagnose various pediatric pathologies. Ultrasonography appeals to pediatric radiologists because it is a non-invasive, non-irradiant, painless, inexpensive imaging method, which is also practicable at the bedside. In patients with bone diseases, this technique has also proved to be a highly effective tool for assessing congenital disorders such as hip dysplasia, infectious processes such as sub-periostal abscesses, inflammatory diseases such as chronic arthritis, and even traumatic events such as ankle sprain. However, with standard devices, this method of examination is not suitable for diagnostic purposes or for monitoring bone tumors, with which other methods are more effective, although they have several disadvantages (cost, relatively low availability, and the irradiation and sedation risks involved).

Many authors have dealt with the ultrasonic imaging of bones. Their main aim has usually been to assess the thickness of the diaphysis [1] and to estimate the speed of sound of a wave crossing the structure [2]. Our group has been focusing on the cross-sectional radial imaging process, using ultrasonic computed tomography. Although this method is known to provide a potentially valuable means of imaging objects with similar acoustical impedance, problems arise when it is proposed to obtain quantitative tomograms of more highly contrasted media (such as hard bone tissues). Finding solutions involves either using non-linear schemes and/or performing extensive studies to reduce the initial approximations. In this paper, we recall the advantages and limitations of ultrasonic computed tomography methods when dealing with highly contrasted scatterers, using a high-order iterative procedure method.

The performances and limitations of this method were then tested using an ultrasonic scanner on a single child fibula, with an adjacent thighbone and geometrical phantom mimicking child bone. The results obtained are promising and suggest that the geometrical and acoustical characteristics of children's bones can be efficiently determined using this ultrasonic computed tomography method.

II. ULTRASONIC COMPUTED TOMOGRAPHY - UCT

The aim of UCT is to perform scattered ultrasonic measurements to build a spatial picture of the distribution of some geometrical and physical parameters of an object. These measurements are carried out using variably densely spaced sets of transmitter and receiver positions and various interrogating wave frequencies. We are then faced with both a forward scattering problem, i.e., predicting the pressure field when the scattering medium and the incident field are assumed to be known, and the inverse scattering problem, i.e., retrieving the parameters of the medium from the incident and scattered fields measured. This inverse problem is non-linear and ill posed, and there generally exists no single solution. It is necessary to find a means of eliminating those solutions, which do not correspond to reality.

Basic UCT principles have by now been clearly established in the case of weakly varying media such as low-contrast structures, i.e. almost homogeneous media [3, 4]. A constant reference medium can therefore be chosen (i.e., approximations will be made with a constant background). The scattering problem can be linearized by using the first-order Born approximation, and if the Green's function of the unperturbed problem (the background) is known, the forward problem can be solved with the Lippmann-Schwinger integral equation, and one method of solving this inverse problem will consist in performing a far field asymptotic development. The "classical" tomographic algorithm will yield the perturbation with respect to the reference problem. This leads to a linear relation between the object function (or contrast function) and the scattered field, particularly in the far-field (2-D or 3-D Fourier transform), which makes it possible in principle to reconstruct the object function in almost real time based on a sufficiently large set of scattering data [5].

In the case of hard biological tissues having larger acoustical impedances than those of the surrounding medium, however, the weak scattering hypothesis is not realistic. If the problem arising for example with the bone thickness imaging is how to identify a water-like cavity within an elastic cylinder, immersed in a water-like fluid, using the low frequency ultrasonic wave (≈ 3 MHz) propagation scheme will be the first solution because the penetration length of the wave increases. In this case, the Born approximation is still satisfied (the wavelengths will be large in comparison with the local acoustical assumptions adopted). The background can be defined in terms of the solid part without any hollow, surrounded by water, and the perturbation, i.e. the object to be reconstructed, namely the cavity. The algorithm of summation of the filtered back-projections can then be used with some signal processing refinements. Despite the artifacts and biases affecting the assessment of the shell thickness, the main result obtained will be a qualitative image of the cavity [6, 7].

However, the first Born approximation will no longer be valid in cases of this kind and it will be necessary to resort to other non-linear strategies. The strategy adopted here involves performing the algebraic inversion of the scattered field, based on the distorted Born iterative (DBI) method, using iterative numerical steps [8]. The DBI method was developed to extend the applicability of the Born approximation to higher orders. Iterations are performed numerically by solving forward and inverse problems at every iteration. With this approach, the medium is modeled without any *a priori* knowledge by performing a simple discretization procedure and using various simulation techniques involving finite differences to calculate the forward diffraction problem. This inversion method consists in updating the medium by performing iterations and requires more computational time and resources, but the algorithm concomitantly determines the appropriate Green's function and the incident field. The previous iteration serves in each case to define the surrounding medium with a variable background, which gives the propagation properties of the medium at the previous iteration. DBI tomography has the advantage of working at a relatively low frequency (≈ 250 kHz). This procedure is based on a mean-square solution obtained via conjugate gradient and Tikhonov regularization methods. The DBI tomography requires only one series of experimental data and five frequencies were used here: 150 kHz, 180 kHz 250 kHz, 300 kHz and 350 kHz. The results thus obtained yield quantitative information about the scatterer, such as speed-of-sound. Very promising results have been obtained previously on synthetic data [9].

III. EXPERIMENTAL DEVICE

The experimental setup used here was designed for performing ultrasonic reflected and diffracted measurements. Six stepping motors sequentially driven by a programmable translator-indexer device fitted with a power multiplexer were used to perform all the movements. The transducers used for the data acquisition were broadband piezo-composite transducers with nominal frequencies of 250 kHz and 3 MHz.

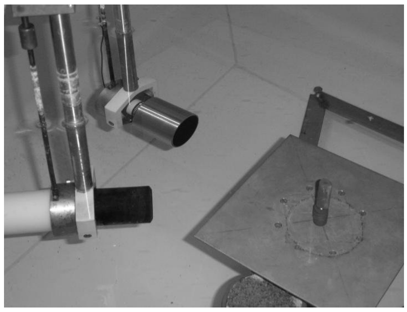

Figure 1. Diffraction/ reflection mode set-up

IV. RESULTS

The validity of UCT was tested on experimental data obtained using the scanner on real child bones and a bone-mimicking phantom.

A. *Real child bone*

The sample was a fresh fibula from a 12-year old child containing no marrow in the inner cavity. The mean cross-section of the bone was 17 ± 2 mm and that of the inner cavity was 6 ± 2 mm. This child bone was surrounding by a tubular-shaped agar gel (30 mm) simulating soft tissues.

Figure 2. Child fibula sample embedded in an agar-gel shell

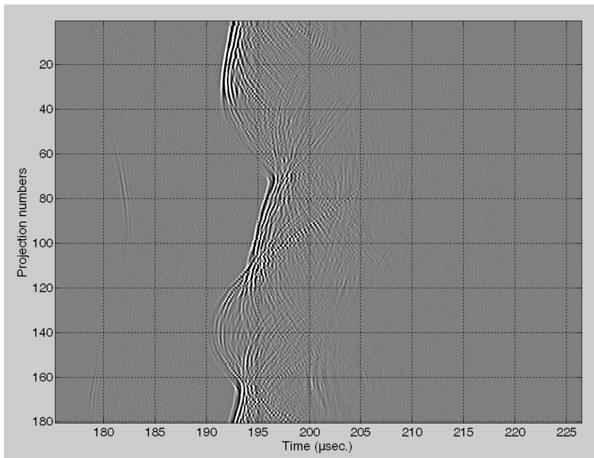

Figure 3. Ultrasonic sinogram ; reflection mode, nominal frequency of 3 MHz, 180 signals / 360°

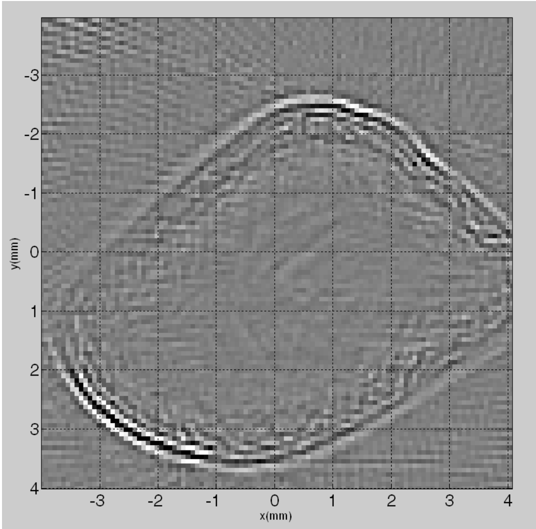

Figure 4. 2-D ultrasonic tomogram, 300 x 300 pixels

Fig. 3 and Fig. 4 show respectively the sinogram and the reconstruction of the child bone surrounded by an agar gel shell. In this case, the UCT procedure adopted was the classical one: a first-order Born approximation and the algorithm for the summation of filtered projections [10] were used. No signal or image processing was performed. The dimensions and the shape of the bone could be readily distinguished on the tomogram, but it was difficult to discriminate between the inner cavity and the cortical area. This is a serious limitation, since the cortical thickness is an index to many bone diseases.

In the second case, the fibula was placed 6 mm from a thighbone so as to create a structure resembling two adjacent bones. This situation required a sufficiently high resolution to be able to detect the gap between the two bones. To check the resolution, we placed a small copper wire (0.6 mm) near the thighbone.

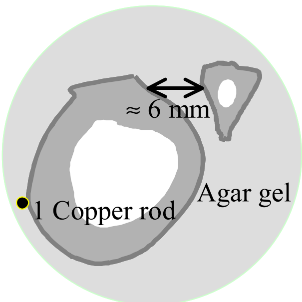

Figure 5. Samples – femur and fibula - with an agar-gel shell

In Fig. 7, it can be seen that the thighbone and the fibula show up particularly clearly. The copper rod is also visible. On the other hand, the zone between the thighbone and the fibula is not clearly imaged and it is impossible to measure the distance between the two bones. This was mainly due to the multiple diffusion occurring between the two objects.

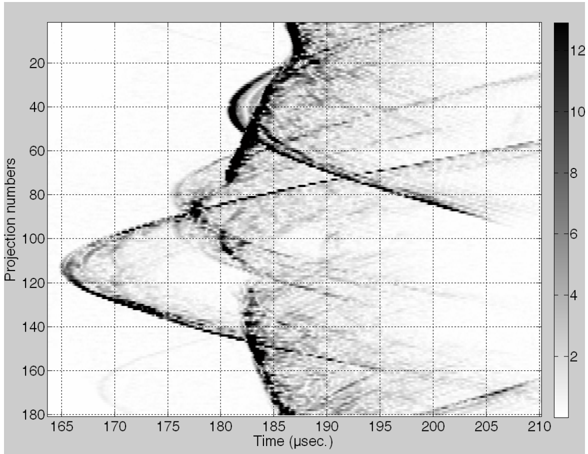

Figure 6. Ultrasonic sinogram; reflection mode, nominal frequency of 3 MHz, 180 signals / 360°

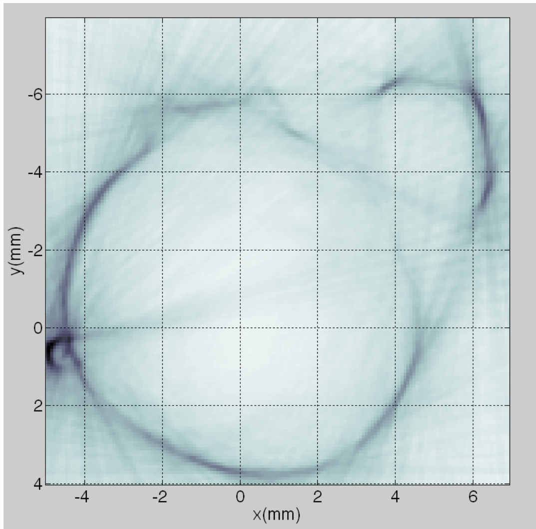

Figure 7. 2D ultrasonic tomography of adjacent fibula and femur bones, 300 x 300 pixels

B. Geometrical phantom

The transmitter used could be rotated step-wise around the target with an angular increment of 5° (72 emission positions). The receiver could be rotated from 40° to 320° with respect to the transmitter position with an angular increment of 10° (29 reception positions at every emission). Fig.8 shows images reconstructed via the DBI inversion method. The experimental sample used here was a non-circular homogeneous isotropic tube made of artificial resin (Neukadur ProtoCast 113™). The density of the resin was $\rho_1 \approx 1150\ kg/m^3$, and the mean velocity of the compressional wave was $c_1 \approx 2400 \pm 50$ m/s. Its maximum internal and external diameters were measured with a caliper and were respectively 6 and 12 mm. It should be noted that in these experiments, differences in the density were observed between the target and the surrounding medium that were not taken into account in the inversion scheme.

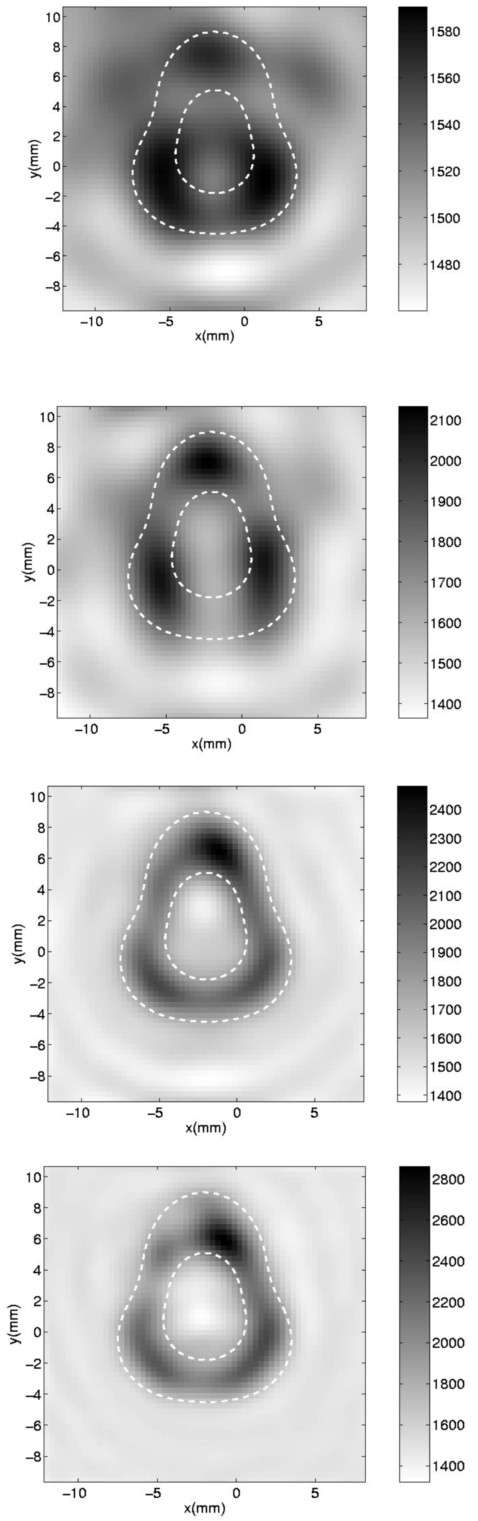

Figure 8. Ultrasonic distorted Born iterative inversion of experimental data obtained on a geometrical phantom mimicking child bone, "Neukadur ProCast" resin, diffraction mode, tomogram 60 x 60 pixels (a) 11 iterations at 150 kHz (b) "a" + 9 iterations at 180 kHz (c) "b" + 14 iterations at 250 kHz (d) "c" + 13 iterations at 300 kHz + 11 iterations at 350 kHz

With then resin of the kind used here to form the outer shell, the density contrast is quite small and can be treated as additional noise when interpreting the experimental data. In addition, it is very difficult to find a high-contrast material having same density as water; the resin adopted here gives the best compromise available. It can be seen here that the resolution and the quality of the modeled contrast improved gradually. The geometry was fairly accurate, whereas the velocity was less accurately assessed, with a relative error of about 5%. This inaccuracy was mainly due to the regularization parameter selected.

CONCLUSION

This paper deals with the two-dimensional imaging of growing bones and a bone-mimicking phantom using ultrasonic computed tomographic methods. The examples tested in this study involve non-canonical homogeneous real and artificial bone shapes. The results obtained are most promising, because the geometrical and physical parameters of the scatterer were accurately reconstructed. Various means of improving this method, especially the DBI tomography, will now be investigated. One important aspect of the inversion scheme used here is the choice of regularization procedure. Another point, which requires further attention, is the constant density hypothesis. Further studies are now required on methods of taking the density contrast into account in the inversion procedure.

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